MEASURING LOWER LIMB LOADING DURING DYNAMIC SPORTING ACTIONS: REDUCING ERRORS FROM SOFT TISSUE INTERFERENCE AND FILTERING

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When determining lower limb loading (LLL) sources of error include the vibration of kinematic marker plates and how the motion and force data are digitally filtered. This study examined LLL during dynamic actions with reduced marker vibration using a tibial plate (TP) and two different levels of filtering severity (15 Hz and 35 Hz cut-offs). The increase of the filter cut-off frequency for the TP led to significantly higher LLL compared to the normally used soft tissue mounted plate. Especially for a cut task, the TP was able to track the high frequency movement components and a higher filter cut-off frequency allowed these to be maintained in the dataset. This suggests that the peak magnitudes of variables linked to the risk of lower limb injury can be underestimated in the literature and there is likely to be a lack of sensitivity in determining 'at risk' individuals.

KEY WORDS: knee moment, tibia, cut-off frequency.

INTRODUCTION: The knee is the most common injured part of the body during running and cutting (Taunton et al., 2002). From cadaver studies it is known that an internal tibial rotation, especially in combination with a knee abduction moment, strains the anterior cruciate knee ligament (Fleming et al., 2001; Markolf et al., 1995). The rate of internal rotation of the tibia during running has been suggested to be a key risk factor for lower limb injury during running. To measure lower leg kinematics during sporting activities a cluster of markers fixed on a plate, is typically mounted on soft tissue, to track the segmental motion (Weinhandl et al., 2016; Willson and Davis, 2008). Unfortunately, for high frequency movements like cutting this limits the accuracy of measuring tibial rotation and as a result it has been very difficult to link transient rotation of the tibia to the risk of knee injury (Benoit et al., 2006; Holden et al., 1997). Relative motion of markers to the underlying bone and a damped oscillation of the soft tissues due to the impact with the ground (Barre et al., 2013; Peters et al., 2010; Reinschmidt, 1996) can influence joint loading estimates. However, filtering of the data can reduce some of those vibrations. We developed a marker plate that is 3D printed to fit exactly the shape of the anterio-medial surface of the distal tibia (with minimal soft tissue between the plate and the underlying bone) which substantially reduces the soft tissue interference. This improved mechanical coupling of the plate to the tibia allows the shank segment motion to be filtered with a cut-off frequency of up to 35 Hz, thereby allowing higher frequency signal to be maintained in the dataset. This study examined the magnitude of LLL using a typical soft tissue marker set/filtering approach compared to simultaneous measures using the individually-moulded tibial plate and processing the data using a higher filter cut-off frequency.

METHODS: 15 male, injury free participants who were regularly involved in dynamic sports like football (age: 23.1 ± 3.5 years; height: 1.79 ± 0.04 m; mass: 73.9 ± 7.9 kg; sport/week: 6.2 ± 2.0 h) performed three different movement tasks: (1) a straight run (4.5 m/s ± 5%), (2) a maximal 90 degree turn with the right leg to the medial side (approach speed: 4.5 m/s ± 5%) and (3) a single leg (right leg) drop landing (40 cm). For averaging purposes, they performed 10 trials for each task (collapsed across five trials in each of two shoe conditions (not part of the current investigation).

For segmental tracking, we used four markers on the shoe and for the shank a marker plate with four markers was attached on the lateral calf using an under wrap and overwrap of stretch bandage. In addition, a tibial plate with three markers was fixed tightly with nonstretch tape on the skin over the anterio-medial surface of the lower third of the tibia (Digby et al., 2005). Kinematics were sampled at 400 Hz with a 12 camera system (Oqus 300, Qualisys, Sweden). For the kinetics a force plate (Kistler, Switzerland) sampled the ground reaction force (GRF) at 1200 Hz. In order to mimic data processing approaches commonly used in the literature the kinematics (soft tissue plate and tibial plate) were filtered with a low pass Butterworth digital filter with a cut-off frequency of 15 Hz. In a second step, the filter cut-off frequency (only tibial plate data) was increased to 35 Hz. Soft tissue plate data filtered at 35 Hz contained substantial vibration errors and was deemed unreliable and unusable. To prevent data processing artefacts the GRF data was filtered in the same way as the kinematic data (15 Hz and 35 Hz) (Bezodis et al., 2013; Kristianslund et al., 2013).

For both data processing approaches the peak knee moments in the frontal plane and peak internal tibial angular velocity (peak IntTibAngVel) were calculated. Peak knee abduction moment (KAbM) during stance was measured for the run. For the cut, all subjects demonstrated a knee adduction moment (KAdM) peak during the first 50ms of stance. This peak was used for the analysis even though a higher knee abduction moment could have occurred afterwards. The frontal plane knee moments for the drop landing task were very subject specific. So for each subject either the KAdM (N = 7) or KAbM (N = 8) was used, depending on which one was higher during the first 100ms of ground contact. For all the analysis Visual3D software (C-Motion, Canada) was used. A one-way ANOVA with a post-hoc multiple comparison (Bonferroni-test) was used to investigate differences between the soft tissue plate, tibial plate 15 Hz and tibial plate 35 Hz (SPSS, IBM; USA).

RESULTS AND DISCUSSION: For the run task a mean value of 1.16 Nm/kg (SD = 0.43) for the knee abduction moment and 14.5° (SD = 4.2) of internal tibial rotation was found (soft tissue plate). This is consistent with the literature (Ferber et al., 2003; Kulmala et al., 2013). The IntTibAngVel of the soft tissue plate (M = 358°/s, SD = 110) is higher than the vast majority of the literature (Bellchamber and van den Bogert, 2000; Mündermann et al., 2003). A reason for that may be the higher sampling rate (400 Hz) used in this study. For the tibial plate 35 Hz, the peak internal tibial rotation (M = 613°/s, SD = 243) is more than four times higher than that reported in the literature (Bellchamber and van den Bogert, 2000; Mündermann et al., 2003). The higher filter cut-off frequency used in this study (15 Hz versus 35 Hz) likely explains most of the difference because the high frequency signal components are retained in the dataset. A good mechanical coupling of the plate to the tibia and using a slightly higher running speed would also elevate the peak angular velocities found in this study. For the cut and drop it is difficult to compare the results to the literature due to different methods.

Generally, the filter cut-off frequency had the largest influence on peak internal angular velocity of the tibia with significant increases compared to all data filtered at 15 Hz (figure 1). The resultant GRF peaks for the cut and drop landing tasks were substantially higher for 35 Hz filtered data (see figure 2). This allows knee moments to be determined using GRF data closely resembles raw collected data and not distorted GRF due to excessive filtering. Therefore, the frontal plane knee moments were also significantly higher for almost all of the 35 Hz filtered datasets. This highlights the importance of higher frequency signal content to in order to measure the magnitude of these specific LLL parameters used commonly in the literature. The internal tibial angular velocity and knee moment curves from 35 Hz filtered data also displayed higher intra-subject variability in the peak values. This may suggest greater sensitivity in the measurement of the loading transients but this needs to be confirmed in future work.

For the 15 Hz filtered data, there were not large differences in the peak values between marker plates. This is partly due to slightly elevated peak values for knee moments and peak tibial internal angular velocities for the soft tissue plate curves due to some degree of oscillation or vibration still present in the data (although the 15 Hz filtering reduced the vibration magnitude). Despite similar 15 Hz filtered data peaks, the tibial plate derived curves reached peak values much earlier (likely due to the greatly improved mechanical coupling to the underlying bone) (see figure 2).

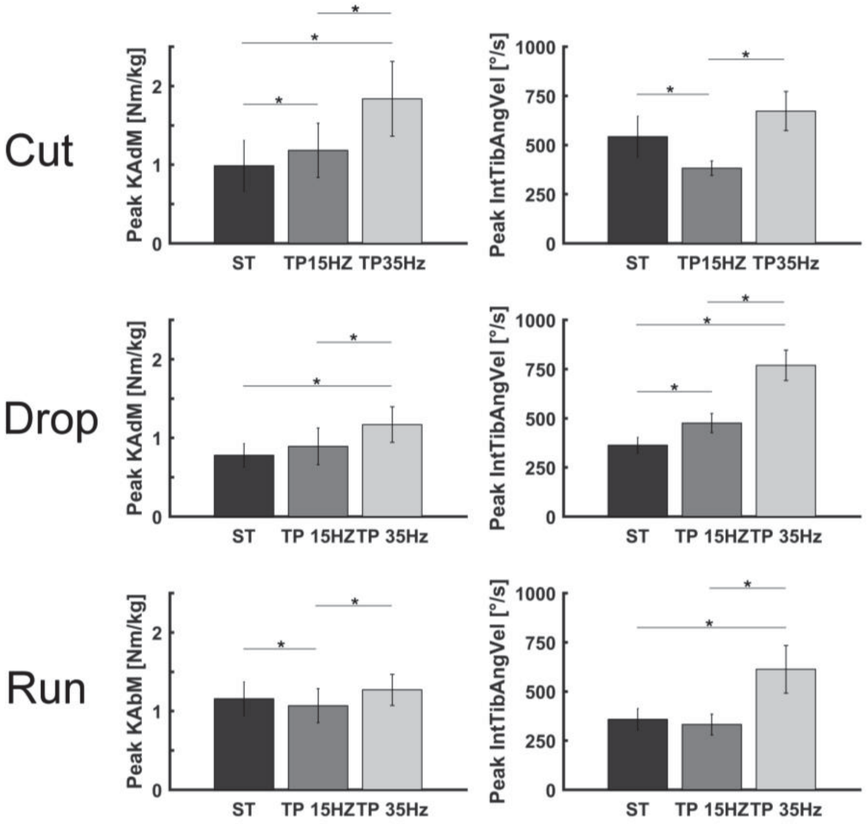

Figure 1: Group means of all three tasks for the soft tissue plate (ST), the tibial plate with a 15 Hz cut-off frequency (TP 15 Hz) and the tibial plate with a 35 Hz cut-off frequency (TP 35 Hz) for the peak knee ab/adduction moment (Peak KAbM and KAdM) and peak internal tibial angular velocity (Peak IntTibAngVel); * significant differences; *p* < .05. The cutting movement tended to demonstrate the highest frontal plane knee moments.

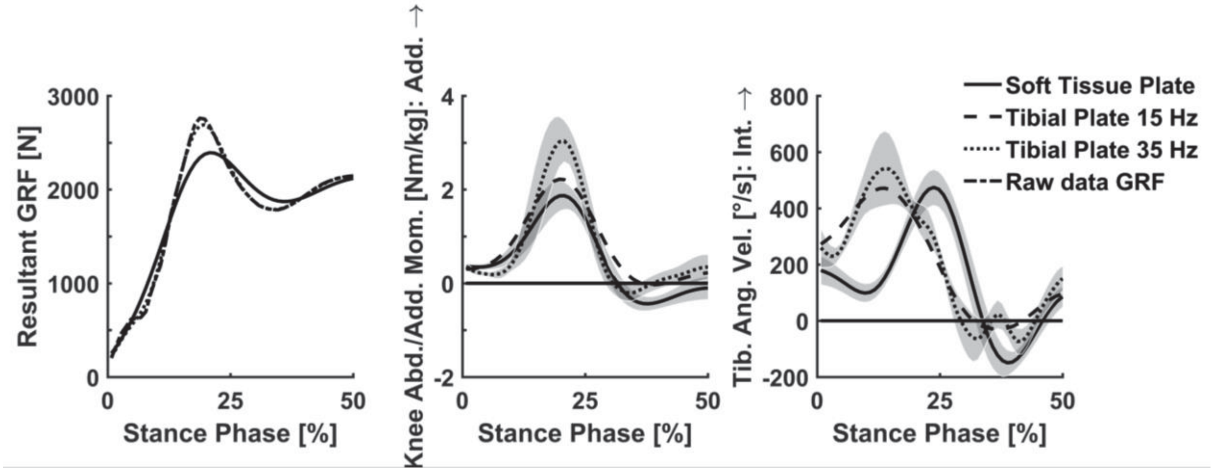

Figure 2: Typical subject curves (mean of 10 trials) for GRF, frontal plane knee moments and internal tibial rotation velocity during the first 50% of pivot foot ground contact of the cut. Notice the higher peaks for the GRF and tibial plate data filtered at 35 Hz and the increased variability in those peaks. GRF data filtered at 35 Hz closely resembled the raw signal whereas there was clear distortion associated with 15 Hz filtering.

Since the cut and drop tasks contain higher frequency signal components compared with the run, it is not surprising that the tibial plate demonstrates larger differences because those curves can be filtered at a higher cut-off frequency and, therefore, higher frequency signal is maintained in the dataset rather than discarded.

CONCLUSION: This study compared a commonly used soft tissue mounted plate on the shank with an individually-moulded plate that fitted over the tibia to estimate lower limb loading variables during dynamic sporting actions. For highly dynamic movements, like turning and landing from a drop, the soft tissue plate underestimated the peak values of the lower limb loadings significantly compared to the tibial plate. It appears that the common practice of reducing marker plate soft tissue vibration errors by using digital filtering (with a relatively low cut-off frequency, 15 Hz) is removing high frequency real signal components in the data (particularly if the same filter is applied to GRF). This leads to an underestimation of the peak magnitudes of specific variables linked to the risk of lower limb injury and possibly a lack of sensitivity in determining ‘at risk’ individuals.

REFERENCES

Barre, A., Thiran, J.-P., Jolles, B.M., Theumann, N., Aminian, K., 2013. Soft tissue artifact assessment during treadmill walking in subjects with total knee arthroplasty. IEEE transactions on bio-medical engineering 60, 3131–3140.

Bellchamber, T., van den Bogert, A., 2000. Contributions of proximal and distal moments to axial tibial rotation during walking and running. Journal of Biomechanics 33, 1397–1403.

Benoit, D.L., Ramsey, D.K., Lamontagne, M., Xu, L., Wretenberg, P., Renstrom, P., 2006. Effect of skin movement artifact on knee kinematics during gait and cutting motions measured in vivo. Gait & Posture 24, 152–164.

Bezodis, N.E., Salo, A.I.T., Trewartha, G., 2013. Excessive fluctuations in knee joint moments during early stance in sprinting are caused by digital filtering procedures. Gait & Posture 38, 653–657.

Ferber, R., McClay Davis, I., Williams III, D.S., 2003. Gender differences in lower extremity mechanics during running. Clinical Biomechanics 18, 350–357.

Fleming, B.C., Renstrom, P.A., Beynnon, B.D., Engstrom, B., Peura, G.D., Badger, G.J., Johnson, R.J., 2001. The effect of weightbearing and external loading on anterior cruciate ligament strain. Journal of Biomechanics 34, 163–170.

Holden, J.P., Orsini, J.A., Siegel, K.L., Kepple, T.M., Gerber, L.H., Stanhope, S.J., 1997. Surface movement errors in shank kinematics and knee kinetics during gait. Gait & Posture 5, 217–227.

Kristianslund, E., Krosshaug, T., van den Bogert, A.J., 2013. Artefacts in measuring joint moments may lead to incorrect clinical conclusions: the nexus between science (biomechanics) and sports injury prevention! British Journal of Sports Medicine 47, 470–473.

Kulmala, J.-P., Avela, J., Pasanen, K., Parkkari, J., 2013. Forefoot strikers exhibit lower running-induced knee loading than rearfoot strikers. Medicine and science in sports and exercise 45, 2306–2313.

Markolf, K.L., Burchfield, D.M., Shapiro, M.M., Shepard, M.F., Finerman, G.A., Slauterbeck, J.L., 1995. Combined knee loading states that generate high anterior cruciate ligament forces. Journal of orthopaedic research : official publication of the Orthopaedic Research Society 13, 930–935.

Mündermann, A., Nigg, B.M., Neil Humble, R., Stefanyshyn, D.J., 2003. Foot orthotics affect lower extremity kinematics and kinetics during running. Clinical Biomechanics 18, 254–262.

Peters, A., Galna, B., Sangeux, M., Morris, M., Baker, R., 2010. Quantification of soft tissue artifact in lower limb human motion analysis: a systematic review. Gait & Posture 31, 1–8.

Reinschmidt, C., 1996. Three-Dimensional Tibiocalcaneal and Tibiofemoral Kinematics During Human Locomotion - Measured with External and Bone Markers. PhD. thesis, The University of Calgary, Calgary.

Taunton, J.E., Ryan, M.B., Clement, D.B., McKenzie, D.C., Lloyd-Smith, D.R., Zumbo, B.D., 2002. A retrospective case-control analysis of 2002 running injuries. British Journal of Sports Medicine 36, 95–101.

Weinhandl, J.T., Irmischer, B.S., Sievert, Z.A., Fontenot, K.C., 2016. Influence of sex and limb dominance on lower extremity joint mechanics during unilateral land-and-cut manoeuvres. Journal of sports sciences, 1–9.

Willson, J.D., Davis, I.S., 2008. Lower extremity mechanics of females with and without patellofemoral pain across activities with progressively greater task demands. Clinical Biomechanics 23, 203–211.